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Individual motion patterns during gait and sit-to-stand contribute to edge-loading risk in metal-on-metal hip resurfacing

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Abstract

The occurrence of pseudotumours (soft tissue masses relating to the hip joint) following metal-on-metal hip resurfacing arthroplasty (MoMHRA) has been associated with higher than normal bearing wear and high serum metal ion levels although both these findings do not necessarily co-exist. The purpose of this study was to examine patient activity patterns and their influence on acetabular component edge-loading in a group of subjects with known serum metal ion levels. Fifteen subjects with MoMHRA (8 male, 7 female) were recruited for motion analysis followed by CT scans. They were divided into three groups based on their serum metal ion levels and the orientation of their acetabular component; well-positioned acetabular component with low metal ions, mal-positioned acetabular component with low metal ions and mal-positioned acetabular component with high ions. A combination of motion analysis, subject-specific modelling (AnyBody Modeling System, Aalborg, Denmark) and CT measurements were used to calculate *dynamically* the contact patch to rim (CPR) distance for each subject during gait and sit-to-stand. The CPR distance for the high ion group was significantly lower ($p < 0.001$) than for the two low ion groups (well-positioned and mal-positioned) during the stance phase of gait (0-60%) and loading phase of sit-to-stand (20-80%). The results of this study, in particular the significant difference between the two mal-positioned groups, suggest that wear of MoMHRA is affected by acetabular cup orientation but also influenced by individual patient activity patterns.

Introduction

The current generation of devices for metal-on-metal hip resurfacing arthroplasty (MoMHRA) were re-introduced because it was hoped that the lower wear rates associated with these devices would reduce the risk of long-term wear induced osteolysis that can cause the failure of metal-on-plastic (MoP) bearings[1]. Well-functioning metal-on-metal (MoM) hip implants show annual linear wear rates of less than 5 µm for the whole articulation per year[2], however, when lubrication is not optimal, the wear rate can dramatically increase[3, 4] (linear wear above 24 µm for the whole bearing per year[5]).

The effects of long-term exposure to metal debris from metal-on-metal wear are largely unknown. There is concern over soft tissue reactions observed in some patients with MoMHRA[6, 7]. These reactions have been referred to by a number of terms such as adverse reaction to metal debris (ARMD[8]), aseptic lymphocytic vasculitis associated lesions (ALVAL[9]), adverse local tissue reaction (ALTR[10]) and pseudotumour[7]. In the current study, the term pseudotumour will be used. Pseudotumours can be extremely destructive, causing one or a number of symptoms such as pain, spontaneous dislocation, pathological fracture and nerve palsy[7]. The outcome of revision of MoMHRA due to pseudotumour can be poor[11].

Pseudotumours in patients with MoMHRA have been shown to be associated with high serum and hip aspirate levels of cobalt (Co) and chromium (Cr); the principal elements of the metal alloy used to manufacture MoMHRA implants[12, 13]. There is evidence that systemic blood metal ion levels are a surrogate measure of wear for metal-on-metal implants[14]. This indicates that pseudotumours are associated with increased levels of wear. Retrieval studies have confirmed that implants revised for pseudotumour have higher wear than implants revised for other reasons[15]. Retrieval studies have also shown that implants revised for pseudotumour are more likely to have experienced edge-loading[16, 17].

Primary edge-loading, sometimes called rim-loading, can occur when the contact between the femoral and acetabular components occurs close to the edge of the acetabular component. It is thought that edge-loading may be a cause of increased wear that leads to high levels of serum Co and Cr ions[13, 18-22]. Low metal-on-metal (MoM) wear rates associated with normal loading conditions have been attributed to the lubrication regime[4] and the formation and interaction of tribochemical reaction layers on the bearing surfaces[23]. Edge-loading is thought to disrupt the lubrication regime between components, leading to a marked increase in localised wear[4]. Under optimal conditions, the diametric mismatch between the femoral and acetabular components forms a "wedge" that allows lubrication entrainment. Contact closer to the edge of the acetabular component prevents the formation of an effective entrainment wedge[24]. Steeply inclined acetabular components are thought to be at greater risk of primary edge-loading. Patients with acetabular cups inclined at angles greater than 55° have been shown to have higher serum levels of Co and Cr ions[19]. In addition to cup orientation, contact stresses near the cup edge demonstrate a relationship with edge radius[25]. During revision of steeply inclined acetabular cups, large amounts of metallosis have been reported[18, 19]. Studies using hip simulators have also demonstrated that wear is increased with increased inclination angle[26, 27].

The risk of pseudotumour is reduced for a cup orientation of 45° (±10°) inclination and 20° (±10°) anteversion[28]. However, this does not represent a true acetabular placement safe zone, as a small number of subjects with "well-placed" components have developed pseudotumours[28-31]. This suggests that there are additional factors involved such as metal hypersensitivity[32], impingement and the individual's activity patterns. Recently, subject-specific activity patterns were shown to have the capacity to reduce edge-loading of steeply inclined acetabular components[33].

Patients with total hip arthroplasty (THA) have been shown to spend the majority of time sitting (44.3%), followed by standing (24.5%) and walking (10.2%)[34]. These data indicate that a considerable number of sit-to-stand (STS) transitions occur daily. It has been estimated that healthy subjects perform STS around 60 times per day[35]. Hip simulator studies have also shown that under simulated start-up and stopping conditions, the wear of MoM implants is increased[36]. Moreover, Wimmer *et al*[37] showed that resting periods increase static friction causing shear force to peak during motion initiation. However, this increase in friction was thought to have no affect on wear as it was attributed to the interaction of carbon layers on the bearing surfaces formed as a result of tribochemical reactions[23, 37]. These layers may come into contact because of a gradual loss of the lubrication layer between MoM components under static loads associated with resting (sitting or lying down)[38]. The time taken to develop adequate lubrication after a prolonged period of rest is unknown.

Wear of articulating orthopaedic implants is a function of the material properties of the bearing surfaces, the contact stress, the type of lubrication regime and the kinematics of the joint. Motion analysis has been used previously to examine the kinematics of the hip joint during gait in order to obtain information on the wear path in THA patients[39-41]. The aim of the current study was to examine the relationship between dynamic patterns of hip loading during functional activity and systemic metal ion levels.

Method: Overview

Langton *et al*[13] introduced the measurement variable of contact patch to rim (CPR) distance, based upon the concept of coverage arc developed by de Haan *et al*[19]. The measurement of CPR made by Langton *et al.* uses an average hip joint reaction force vector and the three dimensional measurement of acetabular orientation from EBRA (Ein-Bild-Roentgen-Analyse)[42] measurements made on static AP pelvic radiographs. The average hip joint reaction force vector used by Langton and colleagues was obtained by averaging measurements taken from four patients in the standing position from the study by Bergmann *et al*[43]. Thus, the CPR measurements made by Langton *et al* are essentially static estimates of the contact patch to rim distance based on generic joint reaction force directions. The CPR value is the distance from the rim to a point at the centre of a contact patch between the femoral and acetabular components. The size and shape of this patch is dependent on a number of factors such as component size, force magnitude and lubrication.

The current study used a combination of motion analysis, subject-specific modelling and CT measurements to calculate *dynamically* the CPR distance for each subject during functional activities of daily living (ADL). The current study then compared the dynamic subject-specific CPR values between patients with well placed and mal-positioned components, and with low and high metal ion levels.

Method: Patients

One hundred and fifty eight MoMHRA patients (201 hips) had their serum metal ion levels measured in an on-going study approved by the local ethics committee. Blood samples were collected in accordance with a previously described protocol[44]. Serum levels of cobalt and chromium were determined using inductively-coupled plasma mass spectrometry (ELAN DRC II, PerkinElmer Life and Analytical Sciences, Shelton, CT, USA) at the Laboratory of Clinical Biology, University Hospital Ghent, Belgium. Fifteen subjects from this cohort participated in the current study (eight females and seven males). These subjects had either a unilateral Birmingham Hip Resurfacing (BHR) (Smith and Nephew, Birmingham, UK) (n=8) or a Conserve Plus (Wright Medical Technology, Memphis, TN, USA) (n=7).

The subjects were divided into three groups (Well-Positioned Low Ions, n=6 [WellPosLow], Mal-Positioned Low Ions, n=5 [MalPosLow], Mal-Positioned High Ions, n=4 [MalPosHigh]) based on the orientation of the acetabular component and their serum metal ion levels (Table 1). Serum metal ion levels were considered high if they exceeded 4.4 µg/l for chromium or 4 µg/l for cobalt[45]. Components were considered mal-positioned if they were outside 45° (±10°) inclination or 20° (±10°) anteversion[28] (Figure 1). It was not possible to find patients in the current overall cohort with Well-Positioned cups and High Ions.

Method: Motion Analysis

Three-dimensional lower limb motion analysis was conducted using a motion analysis laboratory equipped with a 12 camera Vicon Nexus MX system (Oxford Metrics Ltd., Oxford, UK) and 3 force platforms (2 × OR6 AMTI R6-6-1000, 1 × OR6 AMTI R6-7-1000, Advanced Medical Technology Inc., MA, USA). Twenty-five 10 mm diameter spherical retro reflective markers were attached to anatomical landmarks in an established[46] marker configuration, with extra markers on the medial femoral condyles of the implanted limb, on the tibial tuberosities, the medial malleoli, on the distal fifth and first metatarsals and without any greater trochanter markers.

Kinematic data were collected with a sampling rate of 100 Hz and force plate data were collected at 1000 Hz. The subjects' motion was measured during level walking and STS activities. All subjects walked at self-selected normal pace. For the STS activity, the subjects were seated on a bench of height 550 mm with each foot on a force platform. Subjects were then asked to stand with their arms folded across their chests in order to obtain repeatable motions between subjects. All subjects carried out each activity at least six times.

Method: Computed Tomography (CT) Scans

Directly following the motion analysis, computed tomography (CT) scans (Siemens Somatom, Siemens Medical Solutions USA, Inc., NY, USA) of each subject's pelvis and lower limbs were obtained. In order to ensure registration of internal implant positions to the skin-based motion analysis markers, the retro-reflective motion analysis markers were replaced with radio-opaque markers. These multi-modality markers (MM3002, Intermark Medical Innovations, Ltd., Bromley, Kent, UK) consisted of a hydrogel component with a medical grade adhesive.

SliceOmatic (Version 4.2 Rev-9b, TomoVision, Virtual Magic Inc., Montreal, Canada) was used to determine the 3D coordinates of multi-modality markers, the anatomical pelvic landmarks, the MoMHRA prosthesis components and the centre of the hip joint within the CT coordinate system.

Method: Musculoskeletal Modelling

Each subject was modelled performing gait and STS activities in the AnyBody Modeling System (v.5.0) (AnyBody Technology, A/S, Denmark). The model incorporated subject-specific hip joint centres (HJC) calculated for the implanted side by selecting six points from the CT scan on the edge of the acetabular component and fitting a plane to the open face of the component. The average centre of three circles fitted through the six points was projected out from the acetabular component to a distance determined by each patient's component size and component coverage angle. This point represented the centre of the congruent spheres that represent both the outer surface of the femoral component and the inner surface of the acetabular component. All calculations were, therefore, carried out under the assumption that no further separation, other than clearance, occurred between the components during activity. All Conserve Plus implants were modelled with an acetabular component with a coverage angle of 170º and a diametrical clearance of 173 µm[18]. The coverage angle for BHR acetabular components was dependent on the size of the implant and varied from 159.1° to 166.2º (Board & Walter, BHS 2009). The diametrical clearance used for the BHR was 271 µm[18].

On the unimplanted side, the HJC was taken as the centre of a sphere fitted to the femoral head. Subject-specific hip joint contact forces were calculated for each frame of motion captured data. The musculoskeletal model used a three-stage procedure. Firstly, the patient-specific joint kinematics were estimated based on a stick-figure model constructed from the standing reference frame and the estimated HJCs. Secondly, the Twente Lower Extremity Model (TLEM)[47] implemented in the AnyBody Managed Model Repository v.1.2 was non-linearly morphed using Radial Basis Functions (RBF)[48] to match the segment lengths and joint parameters of the stick-figure model. Inverse dynamic analysis was performed for the morphed TLEM model with the measured ground reaction forces as external loads and polynomial muscle recruitment criterion of power 3 to estimate muscle and joint contact forces (Figure 2)[47].

The calculated hip joint contact forces for all fifteen subjects were normalized by body weight (BW), averaged and plotted against the hip contact forces published by Bergmann *et al*[43] for comparison.

Method: Contact Patch to Rim (CPR) Distance

An acetabular coordinate system (ACS) was determined for each subject using points defined on the edge of their acetabular component from the CT images. The HJC, offset from the face of the component as a result of the coverage angle, was used as the origin for this coordinate system. The AnyBody hip joint contact forces were then transformed into this coordinate system.

The location of the intersection of the hip contact force with the inner surface of the acetabular component was then calculated by scaling it by the inner acetabular component radius. This point was assumed to be the centre of the contact patch between the two components. A vector in-plane with the hip contact force vector that intersected with the edge of the acetabular component was then found. The angle in radians between these vectors was found by calculating the inverse cosine of the dot product of their unit vectors. The distance from the intersection of the hip contact force vector with the acetabular component and the component's edge was determined by multiplying this angle by the inner radius of the component (Figure 3). This distance was taken as the contact patch to rim distance (CPR)[13]. The CPR distance was calculated for each patient over the stance phase of gait (0-60%) and the loading phase of STS (20-80%). The average CPR distance per percentage activity for each of the three subject groups was calculated and plotted for comparison. All analyses were carried out using custom software developed in Matlab (R2010b, The MathWorks Inc., Natick, MA, USA).

The lowest 10% of CPR distance for the analysis periods of both activities were found for each subject. These values represented the periods when the contact force was closest to the edge for the load bearing phases of both gait and STS. The lowest 10% was used instead of the absolute lowest value to reflect the variation in CPR distance over the analysis periods. The Mann-Whitney U test was used to determine the statistical significance between the three groups (PASW Statistics, version 18.0.0, SPSS Inc, Chicago, USA).

Due to the gender bias in the MalPosHigh group (female(3)/male(1)) as well as the tendency for females to receive smaller components, the lowest 10% of CPR values for gait and STS were grouped according to gender and then according to component size and analysed. Subjects with femoral components equal in size or smaller than 48 mm (diameter) were classified as 'Small', larger components were classified as 'Large'.

Results: Musculoskeletal Modelling

Overall the hip contact forces computed by AnyBody showed strong similarities with the forces recorded by Bergmann *et al*[43] in terms of magnitude (Figures 4 and 5). However, the calculated forces for gait were most comparable to the Bergmann measured forces at the beginning of the stance phase. At the start of the stance phase the calculated force magnitude peaked at 283 %BW while the maximum Bergmann force was 232 %BW. Towards the end of the stance phase the average force magnitude calculated by AnyBody peaked at 429 %BW. The average force magnitude for the same period was 200 % BW for the Bergmann data (Figure 4). Despite this difference there was no statistically significant difference between the AnyBody and Bergmann forces ($p = 0.945$).

The calculated average force magnitude for STS showed strong similarity with the average Bergmann force magnitude. For the average AnyBody force magnitude, the peak force was 177 %BW while the maximum average force magnitude was 190 %BW for the Bergmann data (Figure 5). However, the variation between subjects was high and this led to a statistically significant difference between the AnyBody and Bergmann forces. When the Mann-Whitney U test was repeated to compare only the highest 10% of AnyBody and Bergmann forces there was no statistically significant difference ($p = 0.338$).

Results: Contact Patch to Rim (CPR) Distance

The subjects in the MalPosHigh group had hip contact forces which were closest to the edge of the acetabular component during the stance phase of gait (Figure 6A). Over the whole of the stance phase the average CPR distance was 17.4 mm (SD 2.5 mm, Range 13.7 – 22.2 m) for WellPosLow, 14.9 mm (SD 1.6 mm, Range 12.6 – 17.9 mm) for MalPosLow and 10.3 mm (SD 2.7 mm, 7.1 – 17.1 mm) for MalPosHigh. When each subject's lowest 10% of CPR values were grouped according to cup position and serum metal ion levels there were statistically significant differences between all three groups ($p < 0.001$) (Figure 6B). When the lowest 10% of CPR values for gait were grouped according to gender there was no statistically significant difference ($p = 0.067$) (Figure 7). When the lowest 10% of CPR values for gait were grouped according to component size there was also no statistically significant difference ($p = 0.44$) (Figure 8).

During the loading phase of STS, the mean values of CPR were 20.5 mm (SD 2.3 mm, Range 15.8 – 23.6 mm) for WellPosLow, 19.4 mm (SD 1.4 mm, Range 16.5 – 22.0 mm) for MalPosLow and 17.4 for MalPosHigh (SD 2.3 mm, Range 13.1 – 21.3 mm) (Figure 9A). Again, when the lowest 10% of values for each patient were grouped into WellPosLow, MalPosLow and MalPosHigh, there were statistically significant differences between all groups ($p < 0.001$) (Figure 9B). There were specific statistically significant differences between WellPosLow and MalPosHigh ($p < 0.001$) and MalPosLow and MalPosHigh ($p < 0.001$) but not between WellPosLow and MalPosLow ($p = 0.309$). When the lowest 10% of CPR values were grouped by gender there was a statistically significant difference between males and females ($p = 0.002$) (Figure 10). When the values were grouped by component size there was also a statistically significant difference between large and small components ($p < 0.001$) (Figure 11).

Discussion

Metal ion levels are now accepted as surrogate measures of *in vivo* wear of metal-on-metal hip arthroplasties. High metal ion levels are associated with mal-positioning and elevated wear rates caused by edge-loading[3, 27, 49-52]. However, some patients with mal-positioned components inadvertently avoid high metal ion levels[28, 29]. The reasons for this are unclear. Lubrication is essential for the proper functioning of MoMHRA, and MoMHRA implants exhibit increased wear when the fluid film lubrication is disturbed, which typically occurs under edge loading conditions. Contact closer to the edge of the acetabular component reduces the potential for lubrication entrainment or the formation of an effective entrainment wedge[24]. It is thought that the disruption of lubrication caused by edge-loading prevents the formation of protective tribo-layers at the bearing.[21] Edge loading occurs either when the acetabular component is implanted with a steep orientation (>55°) (primary edge loading) or when impingement at the neck–cup junction leads to contrecoup edge loading on weight bearing (secondary edge loading).

In this study, we examined a group of fifteen MoMHRA subjects, five of whom had mal-positioned acetabular cups but low serum metal ion levels. The risk of edge-loading during gait and STS was assessed by computing the dynamic CPR distance. There were statistically significant differences ($p < 0.001$) in CPR distance between the MalPosLow and MalPosHigh groups for the lowest

10% of values during gait and STS. In These results suggest that the subjects in the MalPosLow group have motion patterns that insulate their acetabular component from elevated wear rates caused by edge-loading. Interpreting the results of this study in this way could also explain why some patients with well-positioned cups demonstrate high serum metal ion levels[29]. The motion patterns that potentially exert this influence over component wear are a result of anatomy and subject-specific kinematics.

The findings in the current study agree with Langton *et al.* who found a significant inverse correlation between (static standing) CPR and serum metal ion levels[13]. Langton and colleagues observed reduced serum cobalt or chromium when the contact patch was more than 10 mm from the rim[13]. Similarly, in the current study, the mean CPR for the MalPosHigh group during the stance phase of gait was 10.3 mm.

Although it has not been proven conclusively, it has been suggested that the incidence of complications associated with MoMHRA are greater for women[24]. However, there is no evidence to suggest that females are more likely to have a mal-positioned acetabular component i.e. a man is as likely to receive a mal-positioned cup as a woman. This indicates that mal-positioning cannot be the only factor that contributes to high wear. During the loading phase of STS it was found that the edge-loading risk was increased for females ($p = 0.002$). It is possible that higher failures observed in women may be, in part, due to kinematic differences between men and women. For example, it has been shown that females exhibited greater external hip adduction and internal rotation along with hip extension moments compared to males after normalizing for body size for all self-selected walking speeds[53]. In the current study it was also found that during the loading phase of STS, the edge-loading risk was increased for subjects with a femoral component equal to or smaller than 48 mm in diameter ($p < 0.001$). Females in this study had a median femoral component diameter of 46 mm whereas males had a median femoral component diameter of 50 mm. Therefore, the tendency of females to receive smaller components could also put them at risk of edge-loading.

The thresholds for levels of serum cobalt and chrome used in this study (4.4 µg/l for chromium or 4 µg/l for cobalt) to delineate “high” or “low” were based on previous work[45, 54]. The most recent (June 2012) medical device alert issued by the Medicines and Healthcare products Regulatory Agency (MHRA) in the UK suggested that patients with whole blood ion levels of greater than 7 µg/l of chromium or cobalt require close surveillance; however, it is likely that patients with levels >4 µg/L for either Cr or Co are at increased risk of prosthesis failure secondary to increased wear[24]. The MHRA state that cross-sectional imaging should carry more weight in decision making processes than blood metal ion levels alone. This increases the difficulty in relating wear and serum metal ion levels of cobalt and chromium, chiefly because the ways the human body deals with metal wear particulate are not fully understood. To date, only one member of the MalPosHigh group, and none of the MalPosLow or WellPosLow, has been revised due to pseudotumour.

The results in this study did not show evidence of the hip contact force passing through the “edge” of the acetabular component for any of the subjects analysed, i.e. 0 mm CPR. However, the calculations carried out were based on the assumption that the hip contact force vector passed through a point which represented the centre of a contact patch between the femoral and acetabular components. The size and shape of this patch is determined by the force magnitude, the size/geometry/material properties of the components, the clearance between the components and the lubrication (or lack thereof). This represents an extremely complex contact condition which was beyond the scope of the current study to model. Furthermore, the point that we defined as the “edge” does not exist as a distinct locus due to presence of a fillet at the inner surface of the acetabular cup. The radius of this fillet varies between component sizes and between manufacturers and information on the radius of this feature is difficult to obtain. However, despite these limitations, using dynamic CPR analysis and serum metal ion levels as a surrogate measure for wear, it could be hypothesized that during the stance phase of gait, a contact patch on the inner surface of the acetabular component, caused by forces within 8 - 10 mm from the ‘edge’, is of sufficient size to induce edge-loading.

There are a number of other limitations to this study that must be taken into account. For example, the mal-positioned with high ions group had only four subjects three of whom were females. This group also had the smallest component sizes. This gender bias did not appear to affect the CPR calculations during gait, but during STS they may have had an influence. The numbers of subjects with well-positioned components that have developed pseudotumours are small[28-31]. We were unable to recruit a well-positioned with high metal ion levels group to further test the hypotheses that edge-loading is in part mediated by patient-specific motion patterns. Another limitation was the lack of specific information on coverage angle for different sizes of Conserve Plus acetabular components. The value of 170° was taken from information provided by Wright, the manufacturer of the Conserve Plus. However, Griffin *et al.*[55] suggest that the coverage angle for the Conserve Plus is in the range of 162° to 165°. This means that CPR may have been overestimated for subjects with a Conserve Plus. Finally, the methodology presented here cannot account for impingement at the neck–cup junction which leads to contrecoup edge loading (secondary edge loading)[24] nor the influence of micro-separation[56]; both phenomena alter the magnitude and direction of the hip contact force vector and elevate wear rates.

The risk of edge-loading is not only an issue for MoMHRA but also other “hard-on-hard” bearing combinations such as ceramic-on-ceramic. Edge-loading with ceramic-on-ceramic devices is associated with “stripe-wear” and “squeaking”[57]. Currently, recommended values for acetabular component position are applied broadly across patient cohorts and implant designs. The results of this study suggest that, in addition to component position, an individual’s motion patterns play an important role in wear mechanisms. The extent to which motion patterns, specific to an individual, influence wear is unclear. However, there is some suggestion that it is tied to gender. The relative path of the load vector produced as a result of activities such as walking and standing from a chair is variable and dependent upon kinematics. Edge-loading not only depends on component position but also on a patient’s activity pattern. This variability may explain how a relatively steeply inclined acetabular component remains unaffected by edge-loading and some patients with mal-positioned cups do not develop a pseudotumour.

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Tables

Table 1. Composition of subject groups

	Well –Positioned	Mal-Positioned	
	Low Ions (n=6)	Low Ions (n=5)	High Ions (n=4)
Gender (M/F)	4/2	3/2	1/3
Age Years	57	43	46
Weight (kg)	73	73	66
Size (mm)	52	49	47
Time Since Surgery (Yrs)	6.8	5.9	6.5
Chromium (µg/l)	1.5	1.7	6.7
Cobalt (µg/l)	1.5	1.8	6.9

Figures

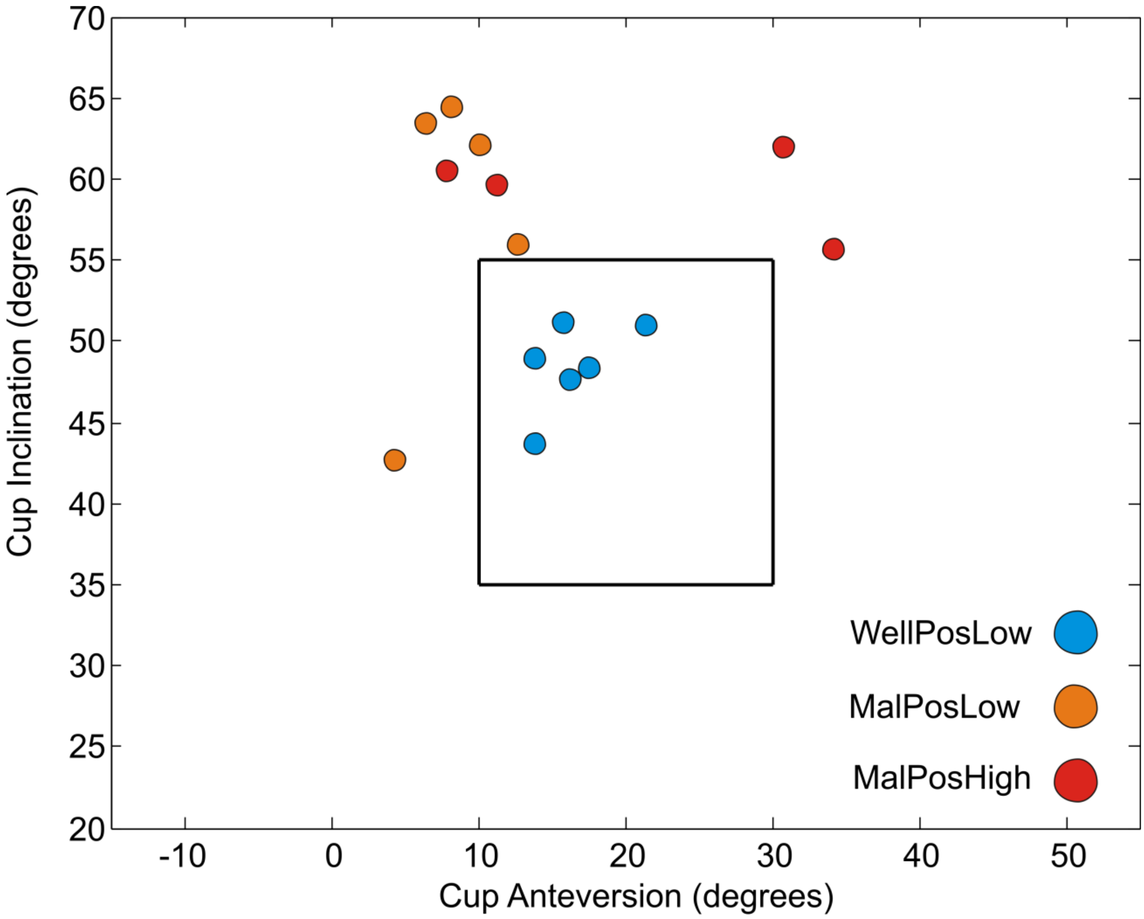


Figure 1. Acetabular cup orientation of all subjects in each group. The box denotes a safe zone for lower risk of pseudotumour identified in a previous publication[28].

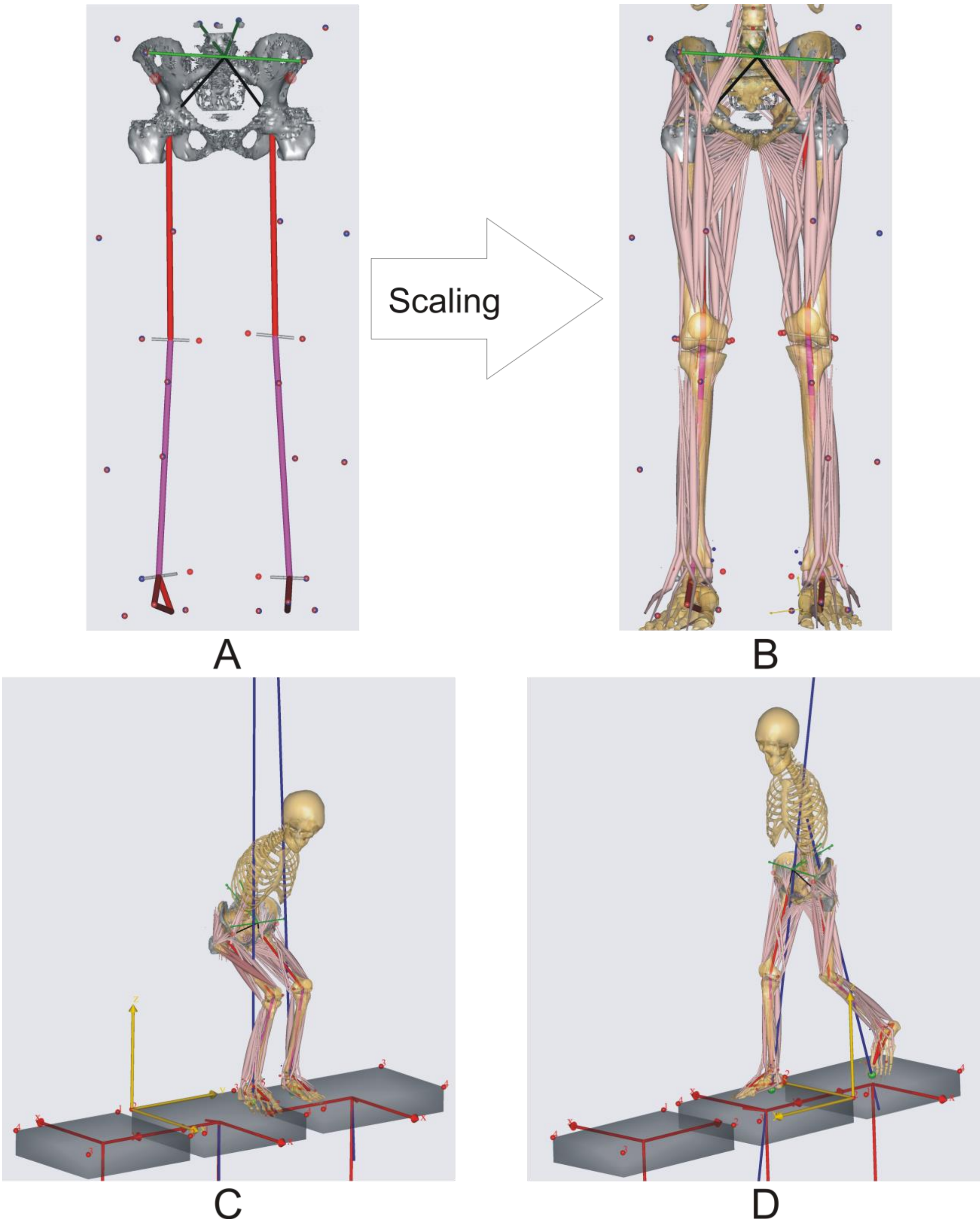


Figure 2. The musculoskeletal modelling workflow for each patient. First, a stick-figure model was derived based on the markers from the standing reference trial and the patient's CT-scan (A). Next, the TLEM musculoskeletal model was nonlinearly morphed to match the stick-figure model (B). Kinematic and inverse dynamic analysis was subsequently performed for STS (C) and gait (D). Additionally, the patients' CT-scan has been overlaid in (A) and (B) to illustrate that both the patient's CT-scan and TLEM pelvis bone match at the points used for morphing on pelvis, i.e. HJCs, ASIS and PSIS.

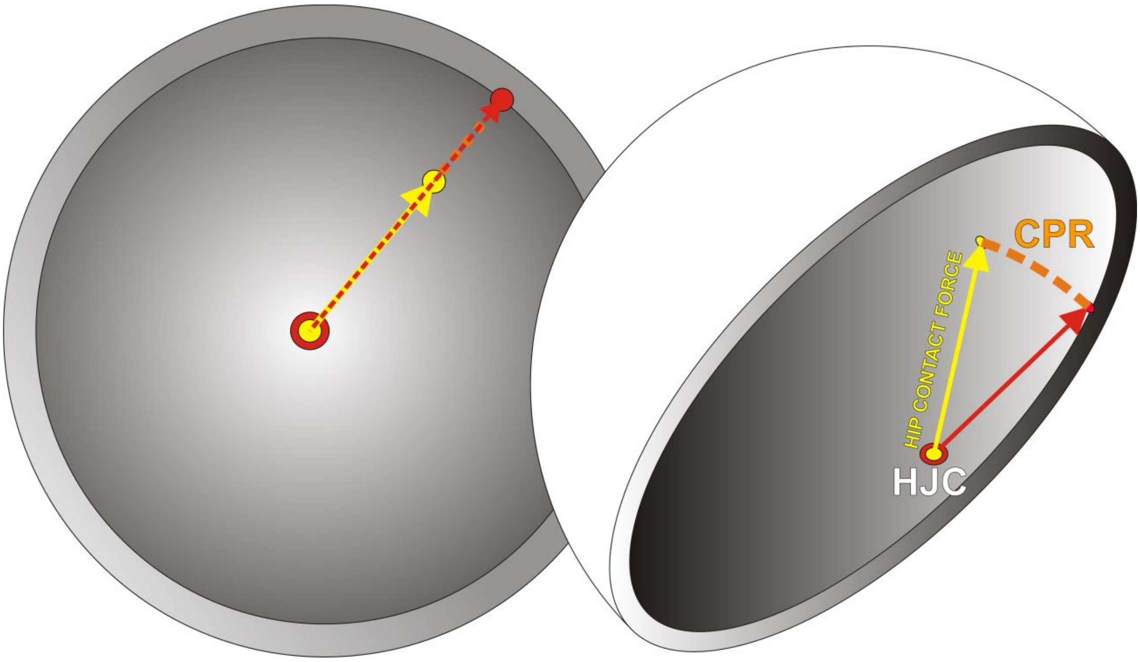


Figure 3. Schematic showing intersection of the hip contact force vector with the acetabular cup and the in-plane edge vector used to find the CPR distance.

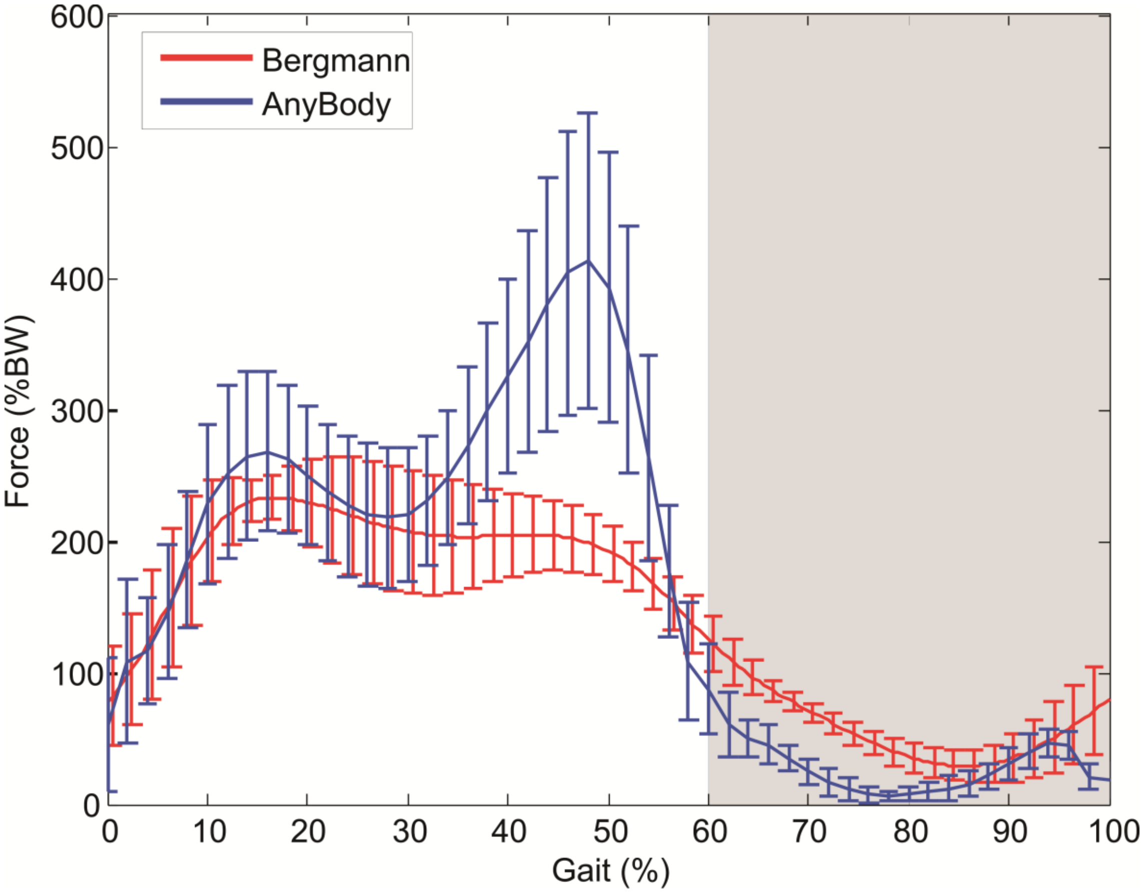


Figure 4. Hip reaction forces estimated by AnyBody and average telemetered forces from instrumented hip prostheses (Bergmann) for gait. Unshaded area is stance phase. There was no statistically significant difference between the AnyBody and Bergmann forces (p = 0.945).

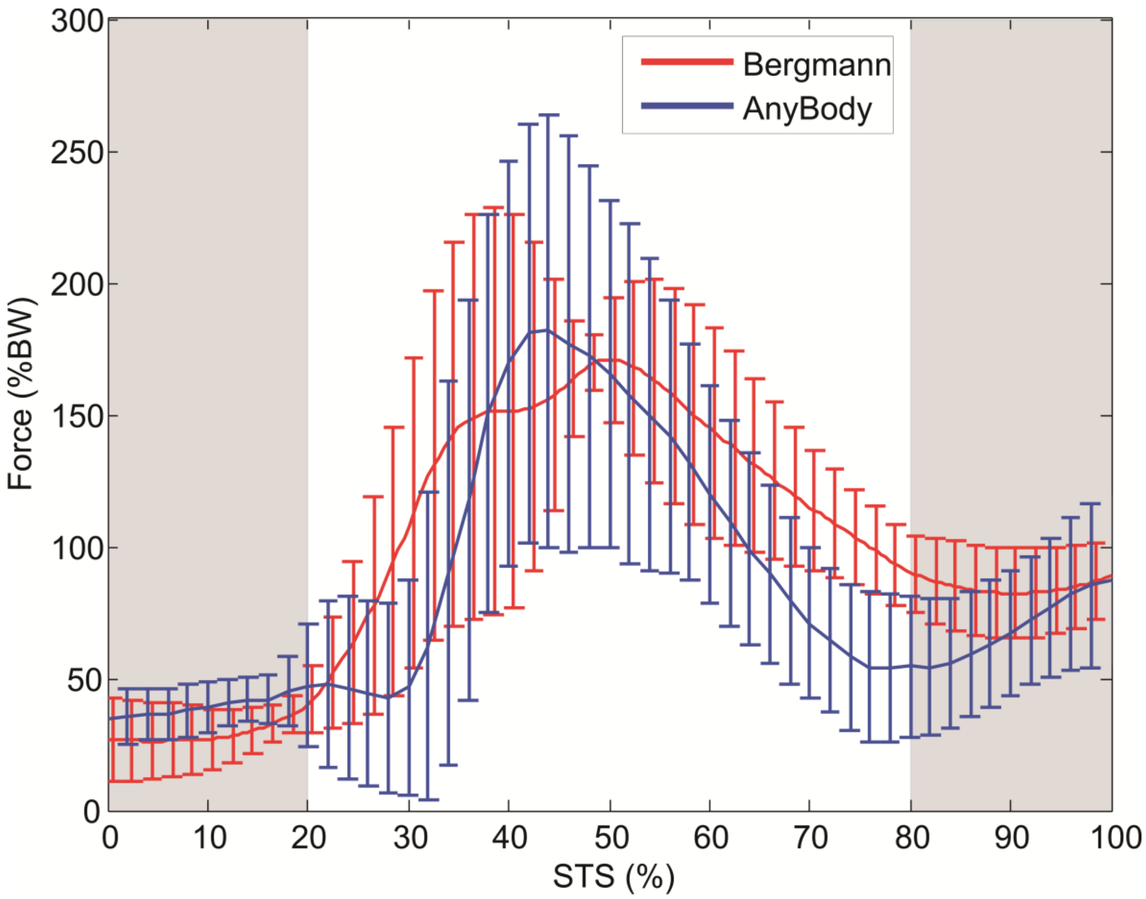


Figure 5. Hip reaction forces estimated by AnyBody and average telemetered forces from instrumented hip prostheses (Bergmann) for sit-to-stand. Unshaded area is loading phase (20 – 80%). There was no statistically significant difference between the highest 10% of AnyBody and Bergmann forces ($p = 0.338$).

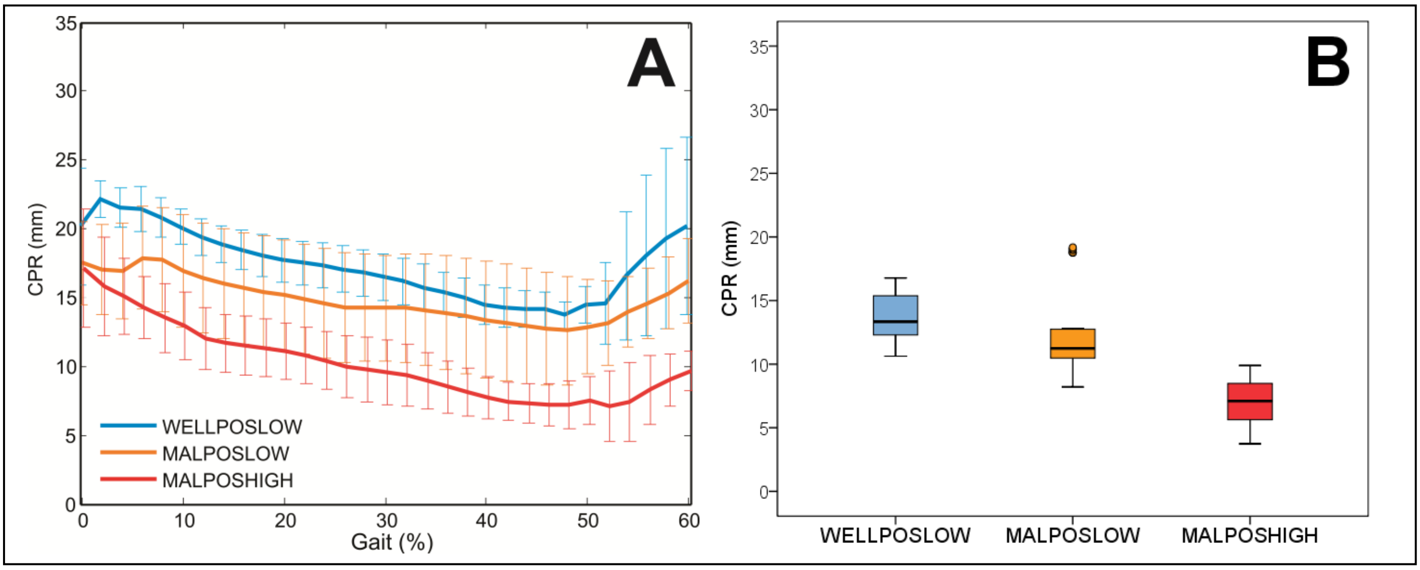


Figure 6(**A**) Average Contact Point to Rim (CPR) with 95% confidence intervals for each group during gait. (**B**) Lowest 10% Contact Point to Rim (CPR) distance during the stance phase of gait for all subject divided by group. There were statistically significant differences between WellPosLow and MalPosHigh ($p < 0.001$), WellPosLow and MalPosLow ($p = 0.003$) and MalPosLow and MalPosHigh ($p < 0.001$).

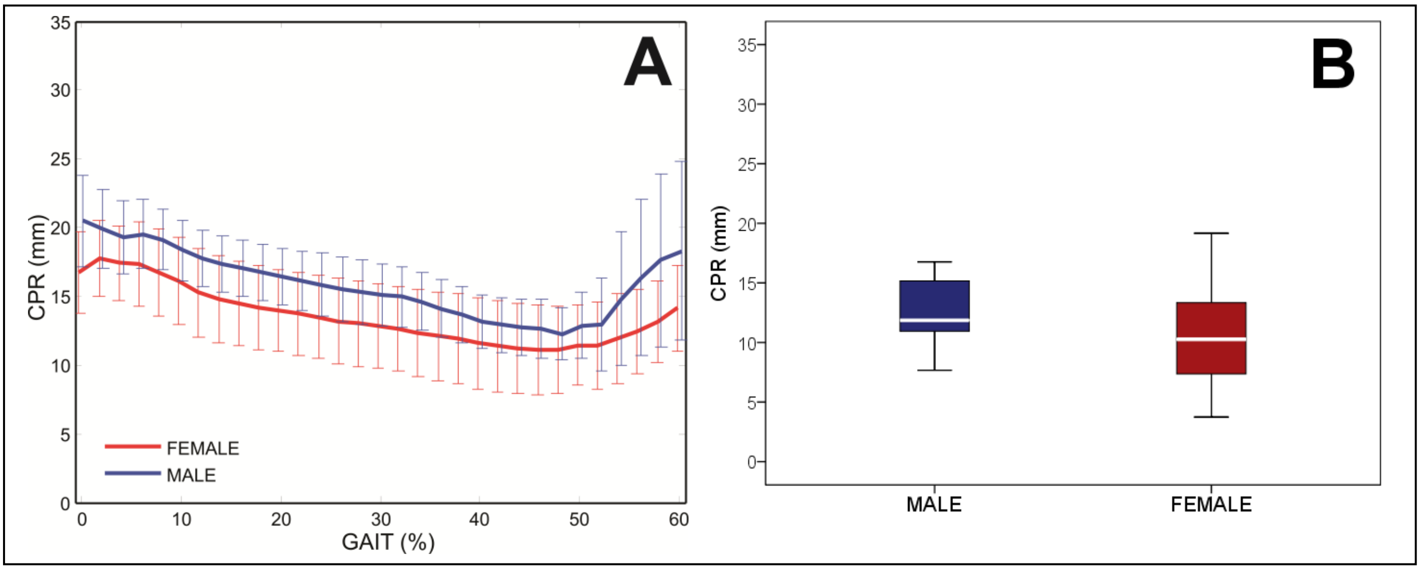


Figure 7(**A**) Average Contact Point to Rim (CPR) with 95% confidence intervals for Males vs. Females during gait. (**B**) Lowest 10% Contact Point to Rim (CPR) distance during the stance phase of gait for males vs. females. There was no statistically significant difference between males and females

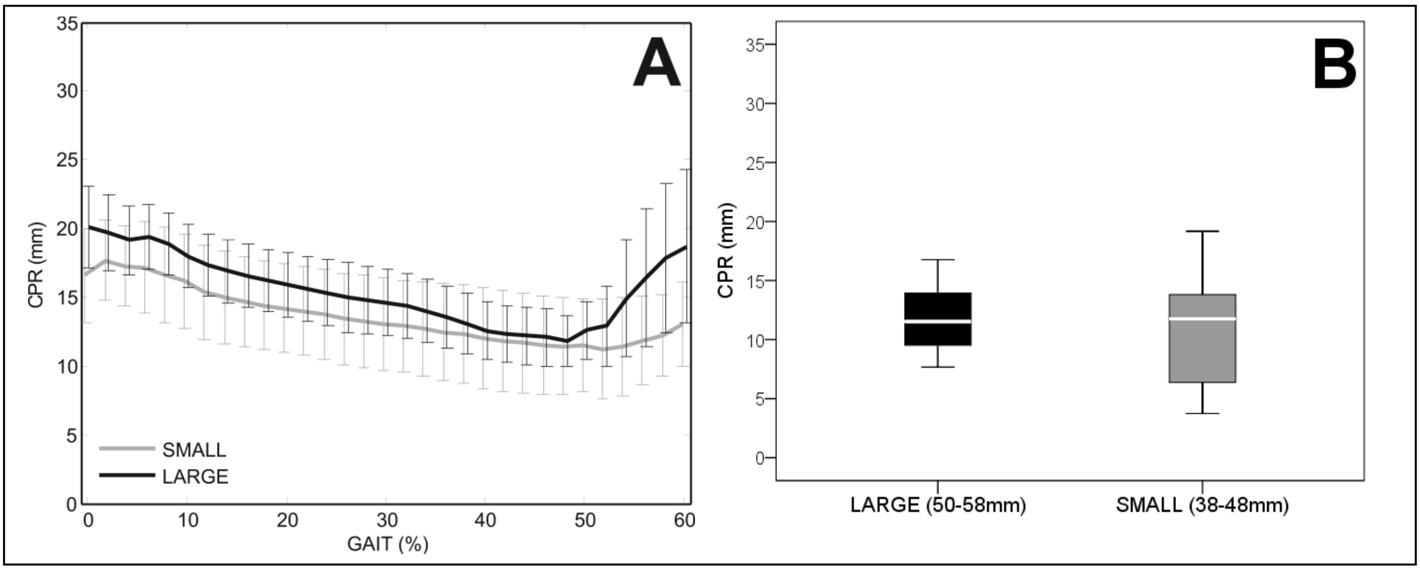


Figure 8(**A**) Average Contact Point to Rim (CPR) with 95% confidence intervals for subjects grouped according to the diameter of their implant. (**B**) Lowest 10% Contact Point to Rim (CPR) distance during the stance phase of gait for subjects with 'Large' or 'Small' components. There was no statistically significant difference between the groups.

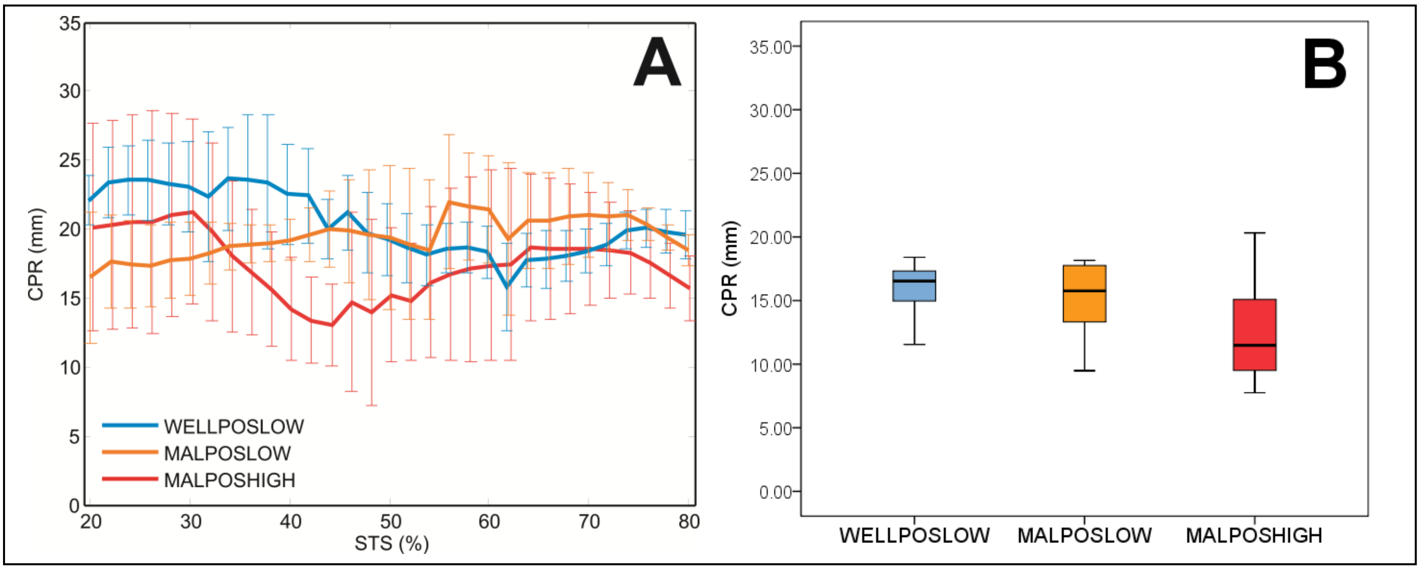


Figure 9(**A**) Average Contact Point to Rim (CPR) with 95% confidence intervals for each group during sit-to-stand. (**B**) Lowest 10% Contact Point to Rim (CPR) distance during the stance phase of gait for all subject divided by group. There were statistically significant differences between WellPosLow and MalPosHigh ($p < 0.001$), WellPosLow and MalPosLow ($p = 0.003$) and MalPosLow and MalPosHigh ($p < 0.001$).

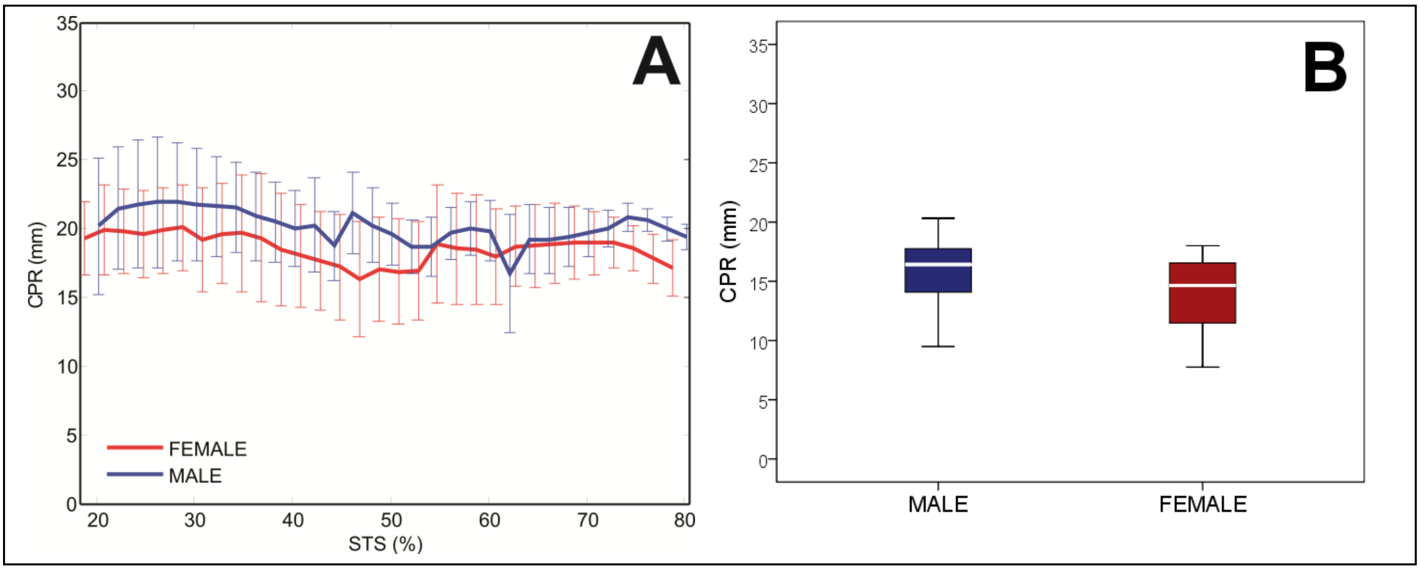
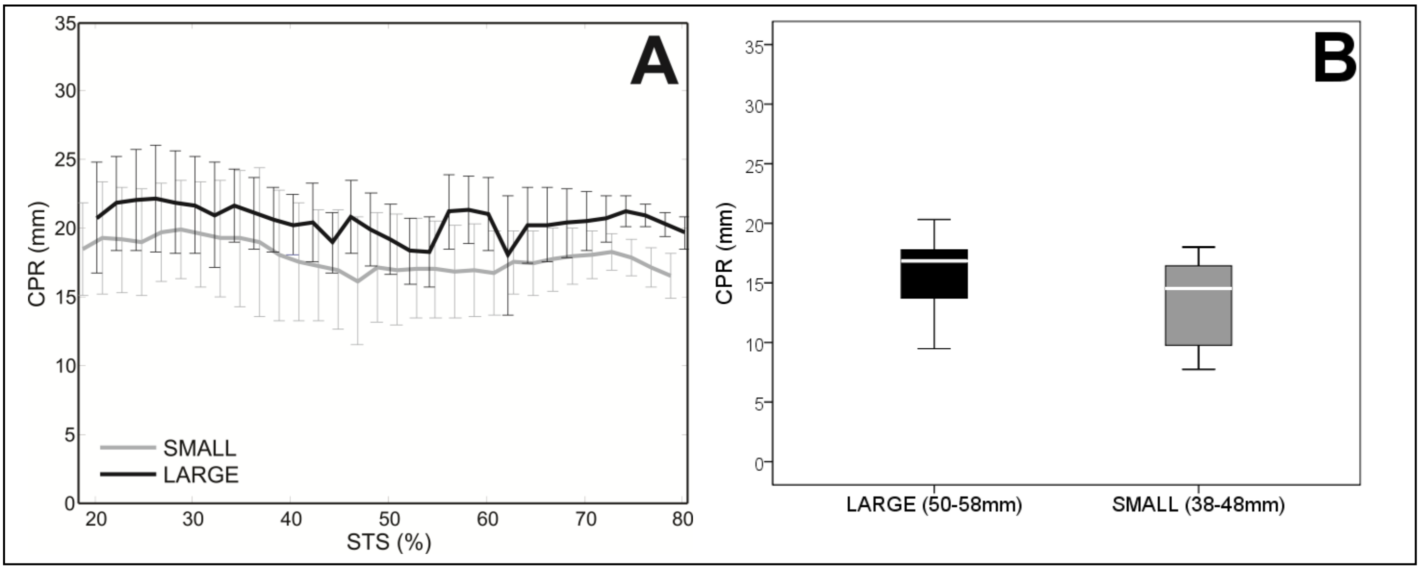


Figure 10(**A**) Average Contact Point to Rim (CPR) with 95% confidence intervals for males vs. females during sit-to-stand. (**B**) Lowest 10% Contact Point to Rim (CPR) distance during the stance phase of gait for males vs. females. There was a statistically significant difference between males and females (p =



0.002).

Figure 11(**A**) Average Contact Point to Rim (CPR) with 95% confidence intervals for subjects grouped according to the diameter of their implant. (**B**) Lowest 10% Contact Point to Rim (CPR) distance during the loading phase of STS for subjects with ‘Large’ or ‘Small’ components. There was a statistically significant difference between the groups ($p < 0.001$).